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THE FEASIBILITY OF A  
FLUIDIC RESPIRATORY FLOWMETER

FINAL REPORT

R-01-09-74

by

Vincent F. Neradka  
Harry C. Bray, Jr.

January, 1974

Prepared under Contract NAS1-11955

by

BOWLES FLUIDICS CORPORATION  
9347 Fraser Avenue  
Silver Spring, Maryland 20910

for

NASA/LANGLEY RESEARCH CENTER  
Hampton, Virginia 23365

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## FOREWORD

This report presents the work performed under Contract NAS1-11955 by the Bowles Fluidics Corporation, Silver Spring, Maryland, during the period September 27, 1972 through January 11, 1974. The work was accomplished under the direction of Mr. Milton Skolaut of the NASA/Langley Research Center.

## ABSTRACT

A study was undertaken to determine the feasibility of adapting a fluidic airspeed sensor for use as a respiratory flowmeter.

A Pulmonary Function Testing Flowmeter was developed which should prove useful for mass screening applications.

The fluidic sensor threshold level was not reduced sufficiently to permit its adaptation to measuring the low respiratory flow rates encountered in many respiratory disorders.

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## SECTION 1.0

### INTRODUCTION

An air flow measuring device has been developed for NASA for V/STOL aircraft applications (Ref. 1, 2). The features which make this sensor attractive for low speed airspeed measurement also offer advantages to the monitoring of respiratory flows. The usual advantages of no moving parts, etc., are supplemented in respiratory monitoring by the device's offering negligible impedance to the inhaled or exhaled flow in a unit which has high speed of response and wide dynamic range. Forced expiratory volume testing as well as continuous monitoring of patients in an intensive care situation is envisioned.

#### General Objectives

The goals of the development program were set forth as follows:

1. Resolution: 5 ml/second
2. Dynamic Range: 0.5 to 8.0 liters/second
3. Frequency Response: DC - > 12 Hz.
4. Linearity: 5% fs
5. Bidirectional sensing capability
6. Determination of relative humidity and temperature sensitivity.

The program was somewhat arbitrarily divided into two phases corresponding to the specified deliverable items, i.e., Item 1 - Pulmonary Function Testing Flowmeter and Item 2 - Intensive Care Unit Flowmeter. This separation allowed for the early fabrication and tests of the pulmonary function testing flowmeter while efforts were continued to optimize the fluidic sensor component.

Each flowmeter employed basically the same sensor which may be likened to a Wheatstone Bridge wherein bridge unbalance is a differential pressure brought about by the stream of air to be measured. The principle of operation is depicted in Figure 1 which shows the power jets each issuing toward its respective receiver, all of which are mounted on a common centerline. Figure 2 illustrates a sensor unit. The receivers measure total pressure recovery, the value of which is altered by the interaction of the jets with the co-flowing and/or counterflowing stream to be measured. This assembly, when installed in a cylinder of known area, provides a volume flow rate measurement.



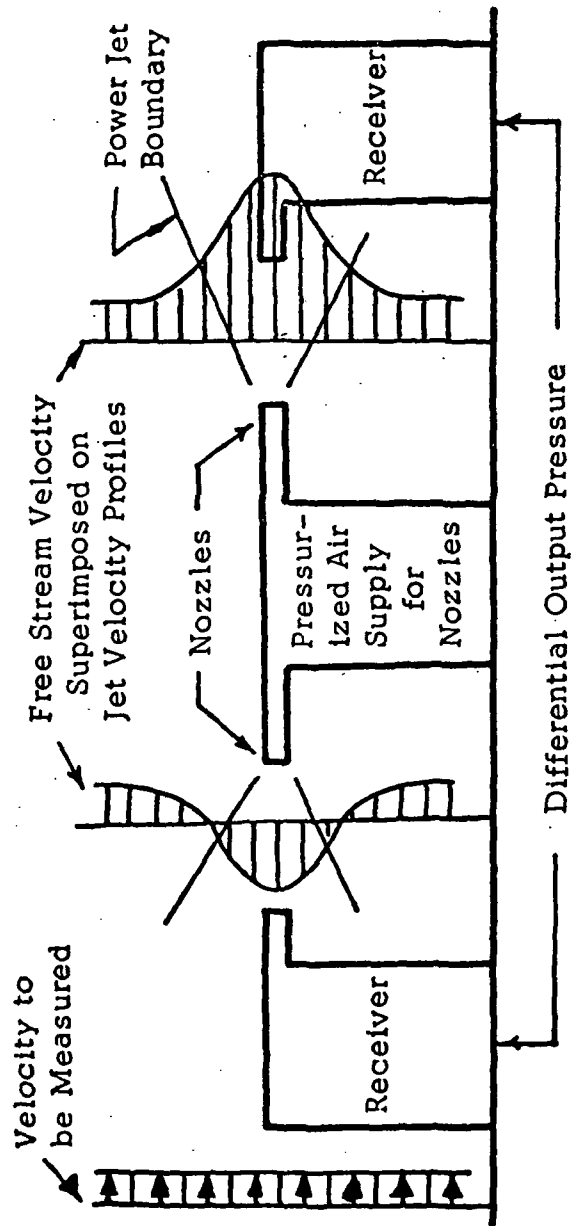


Figure 1. SCHEMATIC OF SENSOR

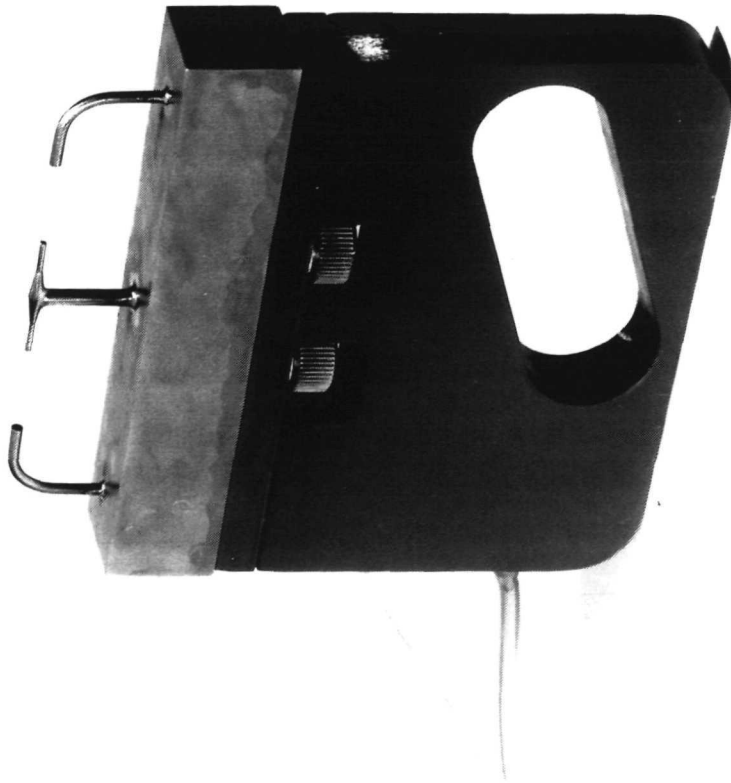


Figure 2. FLUIDIC SENSOR WITH COVER REMOVED

## SECTION 2.0

### PULMONARY FUNCTION TESTING FLOWMETER

Task one of the program required the fabrication of a prototype flowmeter system to determine the feasibility of measuring flow rates and volumes present during pulmonary function testing. Although there is no complete set of tests which satisfy everyone, a series of tests based on a full exhalation have evolved which indicate how disease has altered pulmonary function (Ref. 3). The tests selected result as much from their compatibility to the fluidic flowmeter as to their being basic to pulmonary function testing. These tests are:

1. Forced Vital Capacity, which measures the maximum volumes that can be expelled from the lungs by a forceful effort after a maximal inspiration.
2. Timed Vital Capacity, which measures the volume that is expired by a maximal effort in a specified time.
3. Percentage expired, which is a ratio of timed vital capacity/forced vital capacity.
4. Maximal expiratory flow rate.

For performance of a test, the patient is asked to take a deep breath and expel it as rapidly and as thoroughly as possible through the sensor tube. The four test parameters are recorded and stored electrically. The person administering the test then turns the selector switch from function to function (in any order) to read the data from the 3 digit digital display. A circuit interlock is provided so that subsequent exhalations are ignored. Information is stored until a reset button is depressed in anticipation of the next test. For the purpose of providing a permanent record of the test, an electrical output of the flow rate is available.

A photograph of the system appears in Figure 3.

#### Circuitry

The schematic is shown in Figure 4. The sensor's differential pressure signal is transduced into an output voltage by a strain-gage pressure transducer. This small output voltage is converted to a single-ended output signal by the differential amplifier. A calibrated control adjusts the gain of the amplifier in order to correct for differences between transducers and mouthpieces.

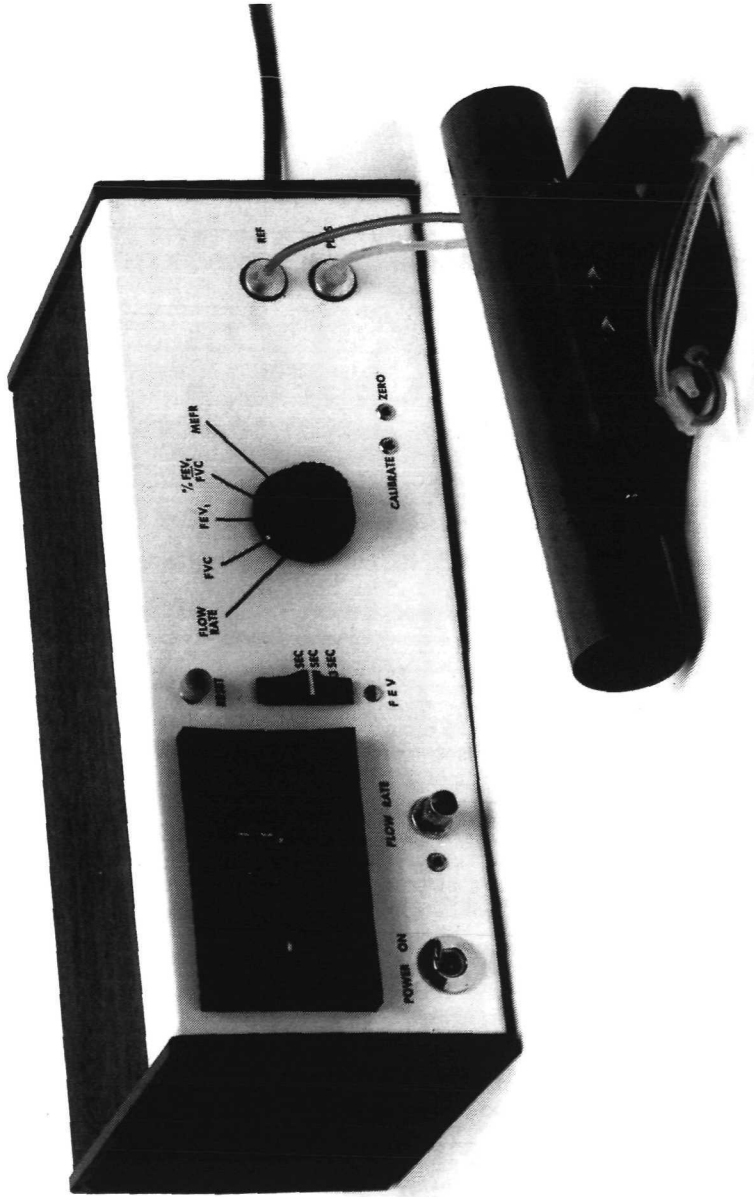


Figure 3. PULMONARY FUNCTION TESTING FLOWMETER SYSTEM



Figure 4. PULMONARY FUNCTION TESTING FLOWMETER CIRCUITRY

The computational circuitry is controlled by the timing and reset logic. Following a reset condition, the timing logic is initiated by a signal from the level comparator. The level comparator indicates an air rate just above the "noise" to reduce errors caused by mouthpiece motion. Upon timing logic initiation, two integrating circuits convert the air rate signal into an air volume signal. After an adjustable pre-determined time, the timing logic stops integration of the Timed integrator. When the level comparator senses that the air flow rate has returned to zero or reversed direction, the timing logic stops integration on the Total integrator. The stored air volume values from the two integrators are processed with a divider circuit that indicates what percent of the total air volume was passed during the first predetermined timing cycle.

A separate circuit, also controlled by the timing and reset logic is a peak detector. This circuit samples the air rate signal and holds the highest signal value obtained during a complete timing cycle. The timing logic is interlocked with a reset to eliminate errors caused by multiple starts. A display selector switch allows the user to display any of the measured or computed parameters on a digital readout. Sample and hold is provided for all outputs except flow rate, and the drift is less than 1% of full scale in 10 minutes. The maximum error from transducer bridge circuit through output is 1% of full scale  $\pm$  1% of value in normal laboratory environments. The system operates on 115 Volts AC 60 Hz.

### Calibration

Figure 5 shows a steady-state calibration of the unit. Apparent on this X-Y recording is the noise superimposed upon the signal. Figure 6 shows the trace of an exhalation as a function of time. On this graph are the values read from the instrument as well as the same data measured from the trace by the use of a polar planimeter. Comparison of the two results shows the correlation to be quite good with the biggest error appearing in the maximum expiratory flow rate parameter. This difference is attributed to noise generated by the fluidic sensor. Additionally, it was determined that the calibration potentiometer was incorrectly adjusted.

When the unit was tested with a Wedge Spirometer as a reference, measurement of the peak flow rate and measurement of very low flow rates resulted in more than acceptable error. Figure 7 shows the trace of an exhalation as a function of time by the Wedge Spirometer reference. Also on the graph are the values read from the instrument. Again, the discrepancy in the low flow rate region is attributed to noise of the fluidic component. The discrepancy in the measurement of peak flow rate is attributed to noise and the maladjusted calibration potentiometer.

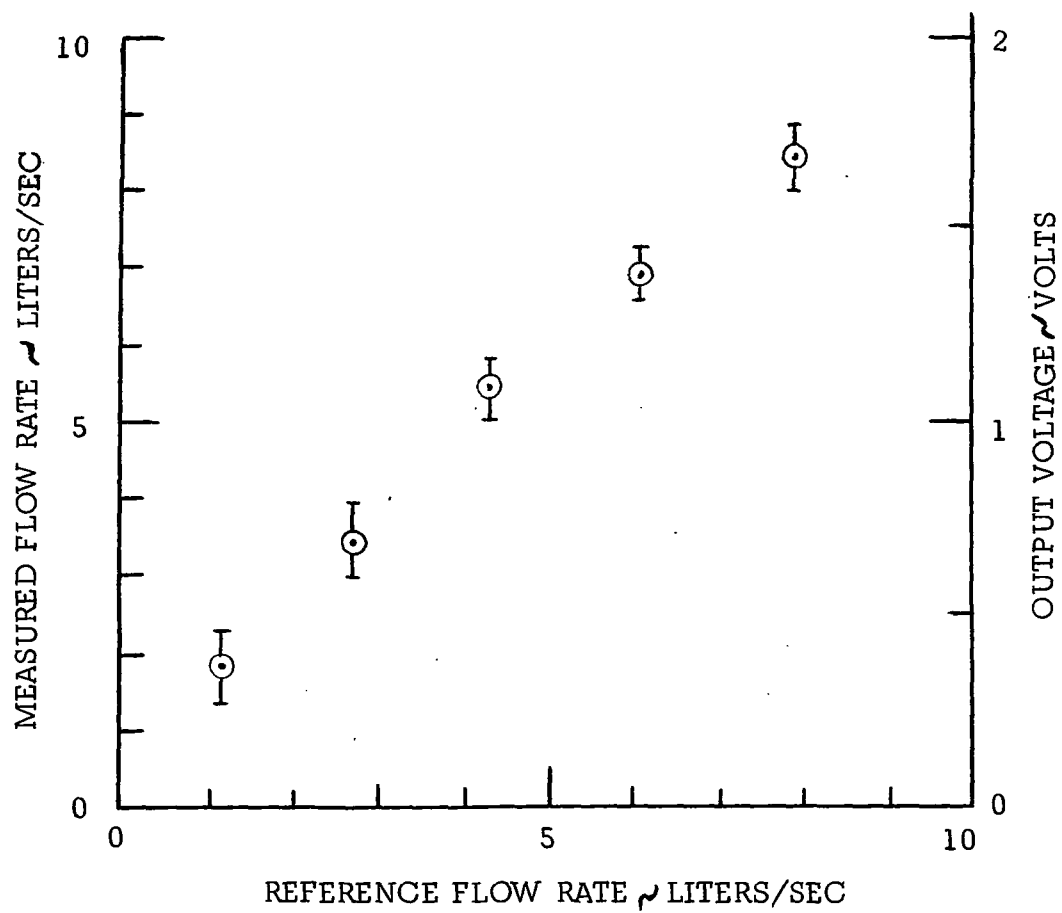


Figure 5. PULMONARY FUNCTION TESTING FLOWMETER  
STEADY STATE CALIBRATION CURVE

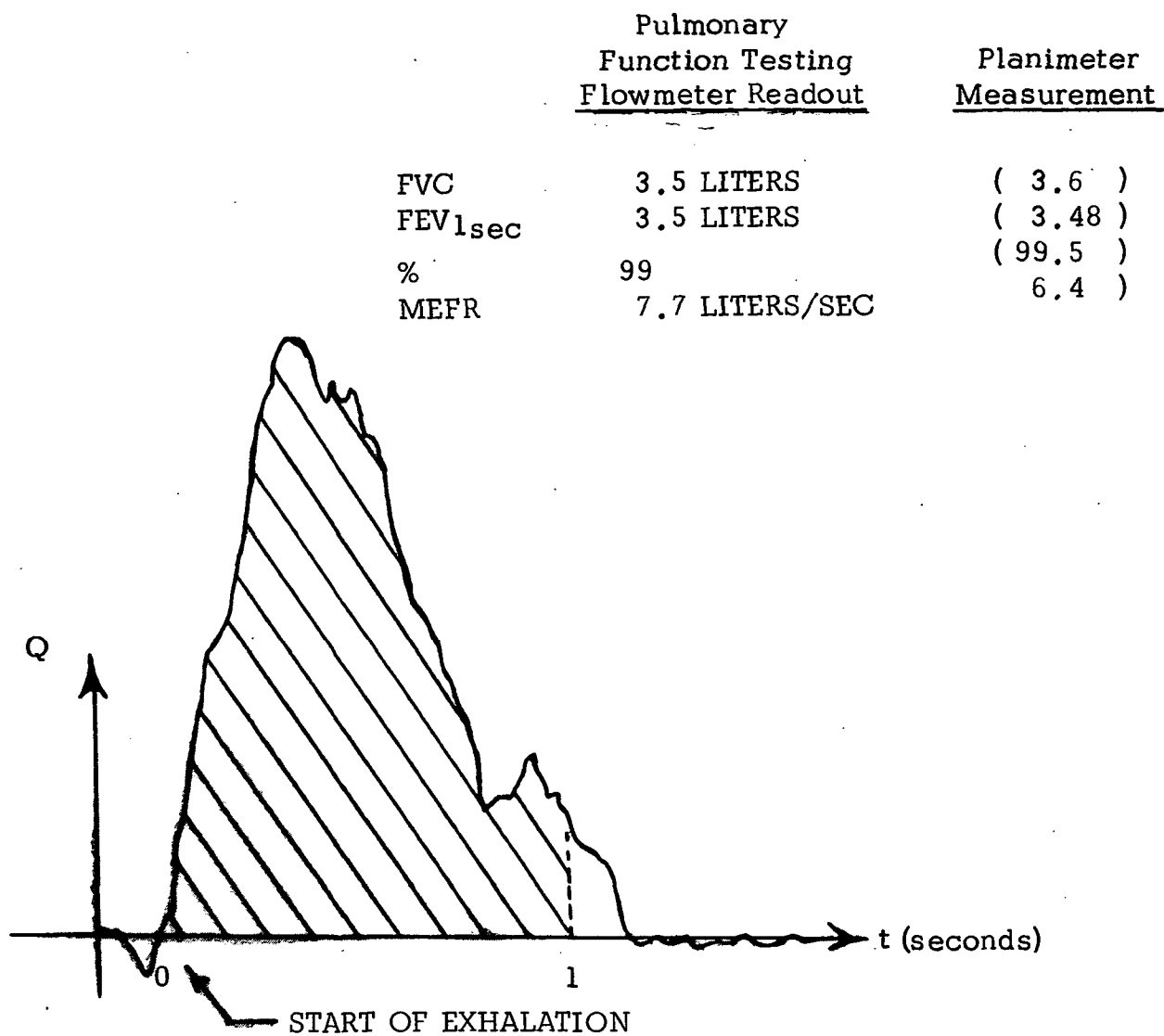


Figure 6. EXHALATION TRACE



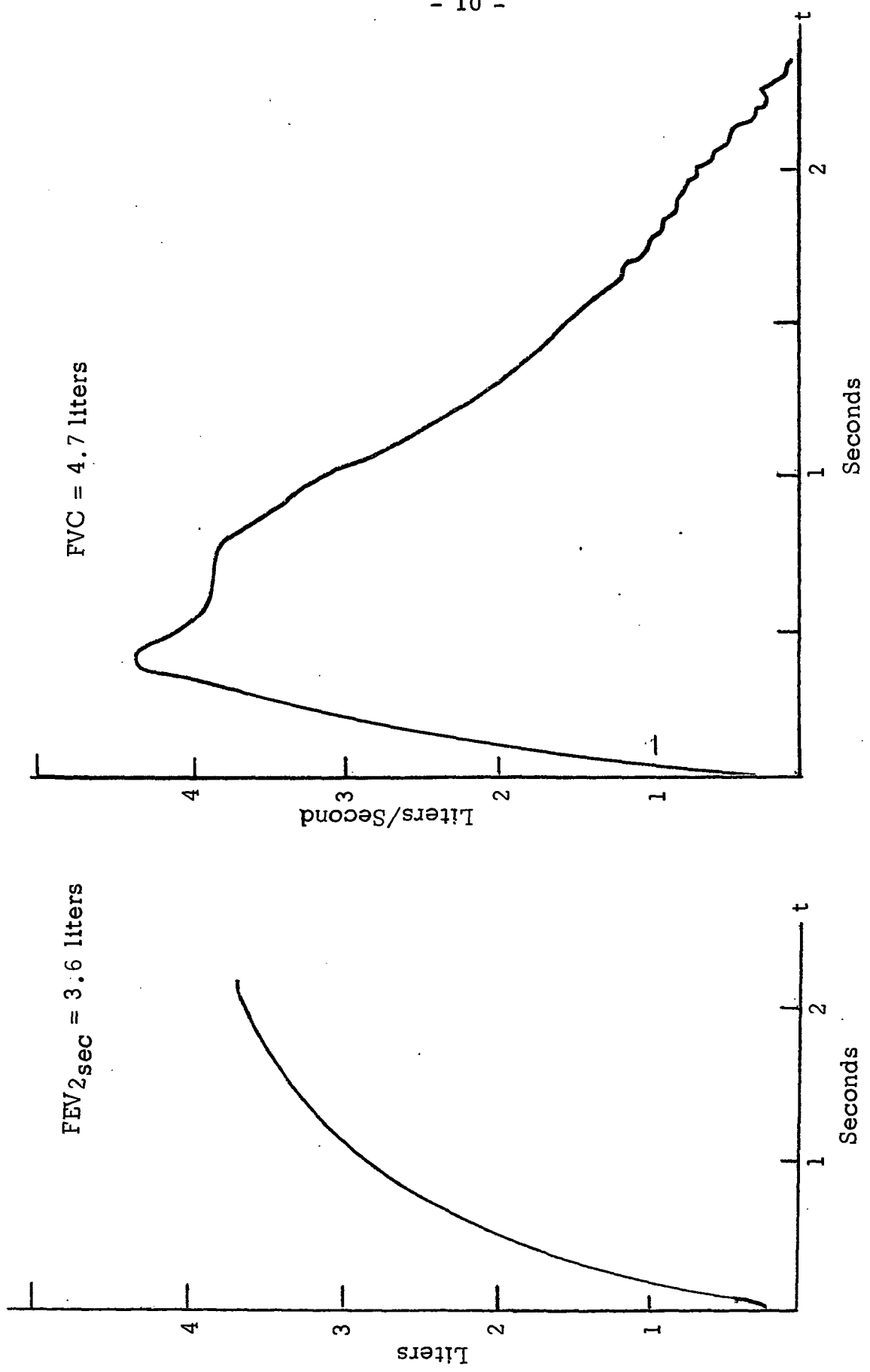


Figure 7. CALIBRATION OF PFTF USING WEDGE SPIROMETER REFERENCE

## SECTION 3.0

### GAS COMPOSITION TESTS

Since the sensor unit will be exposed to moisture-laden air in pulmonary function measurement, its sensitivity to such conditions has been tested. Tests were conducted with dry and saturated air at three temperatures. A nebulizer flow also was introduced at one temperature.

#### Test Set-Up (See Fig. 8)

Laboratory air was supplied as the power supply for the unit. The test gas temperatures were 21°C, 37°C, 41°C.

Saturated gas was provided by mixing steam with laboratory air in a large stilling chamber. The mixture was passed through a heat exchanger which condensed out the excess moisture proportionally, according to whether the temperature was controlled to 21°, 37°, or 41°. This saturated air was immediately passed through the Hastings-Raydist Mass Flowmeter and through the test section of the Fluidic Flowmeter.

Dry gas was provided by admitting dry air to the settling chamber, where it mixed with some laboratory air (which is relatively dry because of the air-conditioning). This mixture was passed through the heat exchanger to warm it to the desired temperature.

Relative humidity and temperature of the air was measured as it entered the mass flowmeter.

In each test a 1 Hz. filter was employed to facilitate measuring the differences brought about by the temperature and humidity and a trace of sensor output differential pressure versus flow rate was obtained.

Test data for saturated air is shown in Figure 9 and data for dry air is shown in Figure 10. In each case, the data indicates there is a decrease in output pressure with increasing temperature.

Figure 11 shows a graph of flowmeter sensitivity variation with temperature of the gas being measured. It is apparent that the variation is approximately linear with a gain of 0.7%/°C and 0.06%/RH.

If it is assumed that the normal operating point is 31°C and 50% RH, a computation of the worse case error indicates that it is in general agreement with the program goal of 5% accuracy.

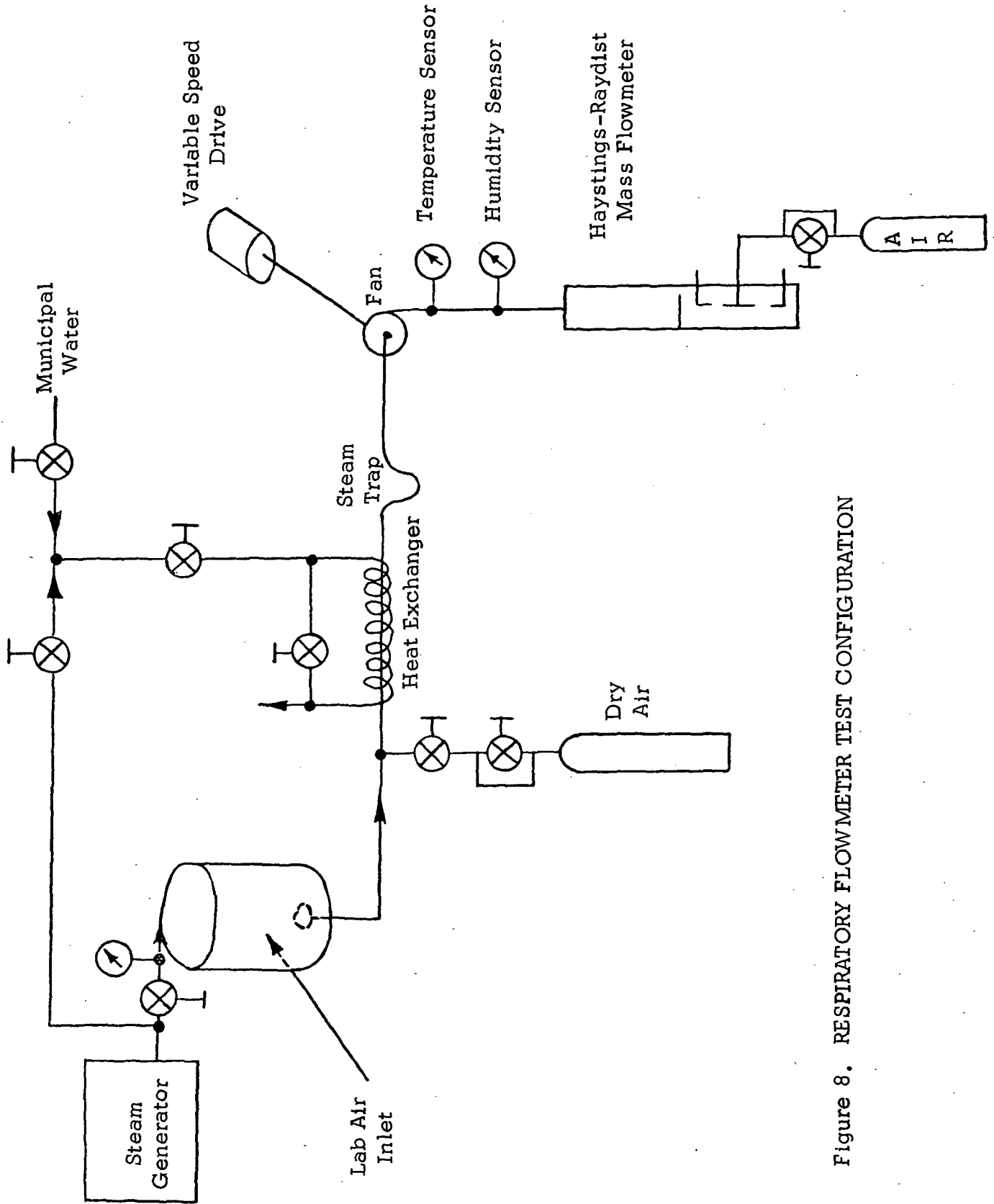


Figure 8. RESPIRATORY FLOWMETER TEST CONFIGURATION

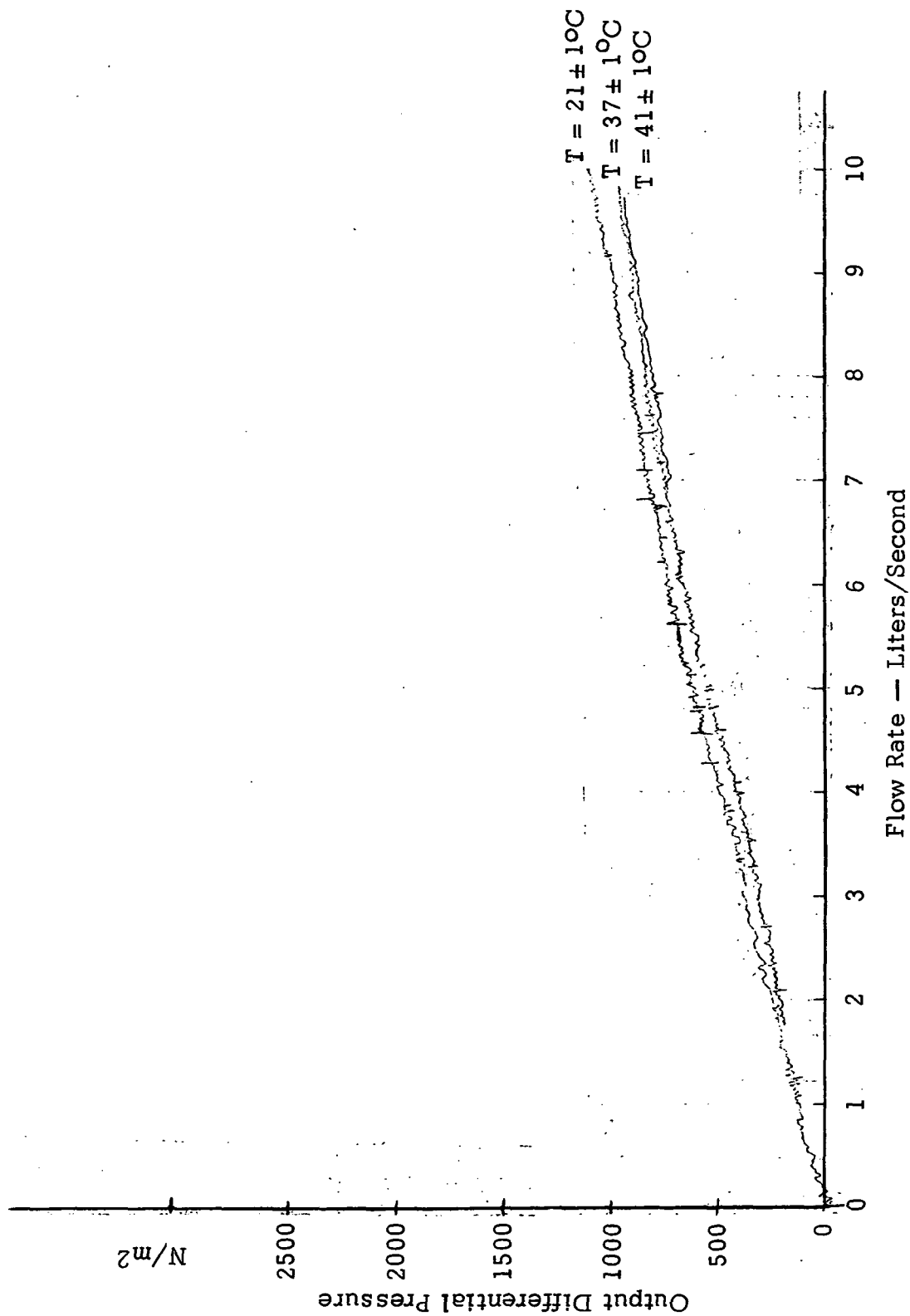


Figure 9. RESPIRATORY FLOWMETER TESTS WITH SATURATED AIR

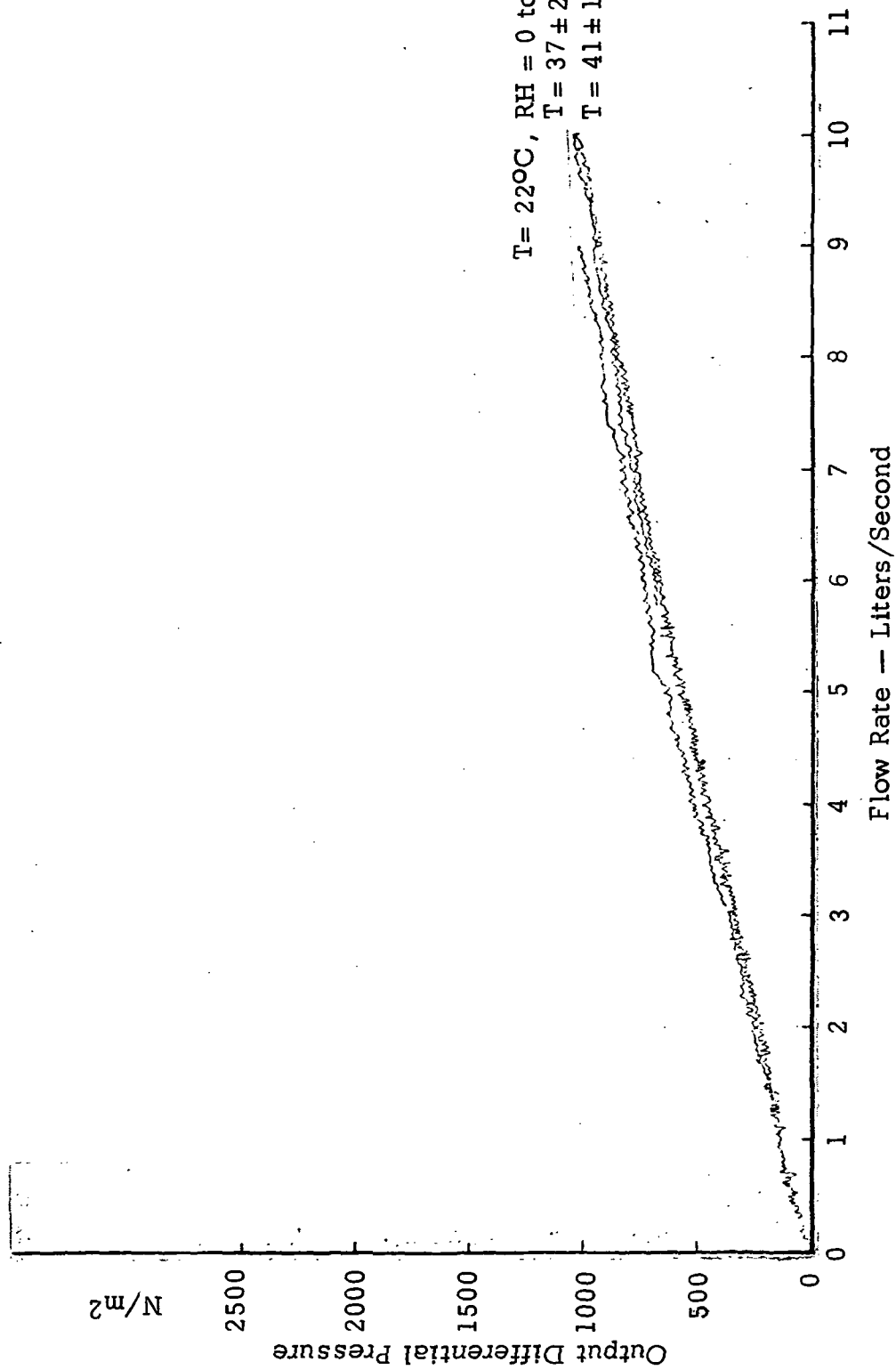


Figure 10. RESPIRATORY FLOWMETER TESTS WITH DRY AIR

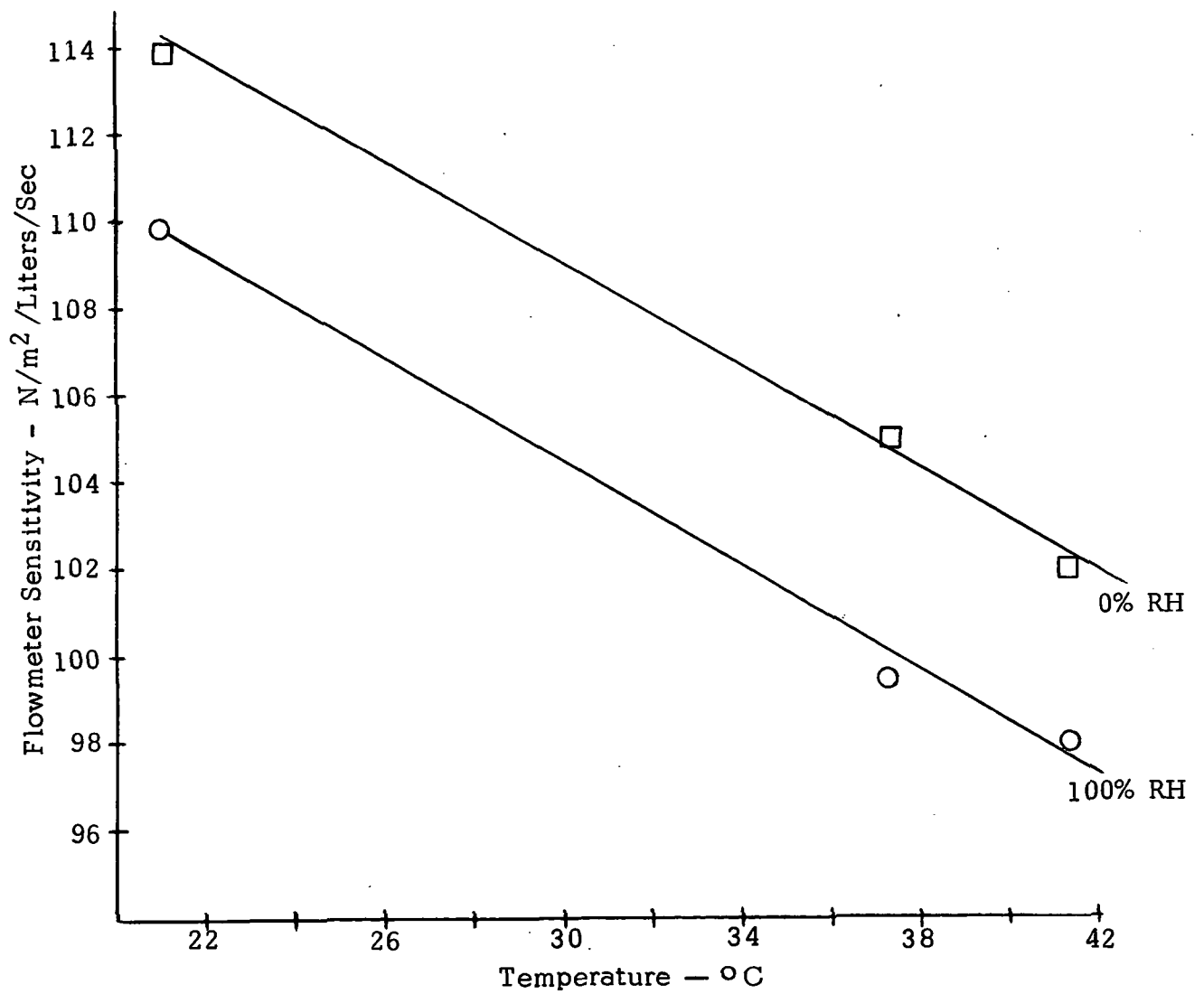


Figure 11. RESPIRATORY FLOWMETER VARIATION WITH TEMPERATURE

A nebulizer flow was introduced downstream of the mass flowmeter to determine the fluidic sensor sensitivity to water droplets contained in the sensed flow. The test was conducted at 37°C with laboratory air. The nebulizer produced a reported water droplet size of less than 25 micron diameter. Test data shown in Figure 12 indicated that there is no significant change in output differential pressure level with the introduction of the nebulizer flow. The error arising from measuring flow before the nebulizer is small due to the minute flow of the nebulizer itself ( $<0.05$  L/min).

## SECTION 4.0

### INTENSIVE CARE UNIT FLOWMETER

A second deliverable prototype item required was an Intensive Care Unit Flowmeter which would be used for continuous patient monitoring. The requirements of this unit differ from the Pulmonary Function Testing Flowmeter requirements primarily in two ways:

1. The electrical circuit requires no computation or digital readout.
2. The flow rate sensitivity is to be improved by reducing the threshold by two orders of magnitude.

A photograph of the Intensive Care System appears in Figure 13.

#### Circuitry

The Intensive Care Unit schematic is shown in Figure 14. The unit operates on 115 Volts AC 60 Hz electrical power.

#### Sensor Development

It became apparent during the early months of the program that the fluidic sensor noise would have to be reduced by at least two orders of magnitude in order to attain the program goal of a 5 milliliters per second threshold.

The noise reduction problem was approached on several fronts. Among the efforts pursued were:

- A) Optimization of power nozzle and receiver design
- B) Optimization of sensor geometry
- C) Optimization of sensor supply pressure
- D) Supply pressure variation

The results of this investigation are summarized below.

A) Power nozzle and receiver design -- Early sensor models were of a square nozzle design due primarily to manufacturing considerations. Pantomilling and dip-brazing were thought to yield the most repeatable results. The square nozzle design provided good linearity and repeatability but at the expense of a high output noise level. Subsequently round nozzles were fabricated by a "lost wax" process. In this process, nickel is electro-deposited onto an aluminum mandrel which is then



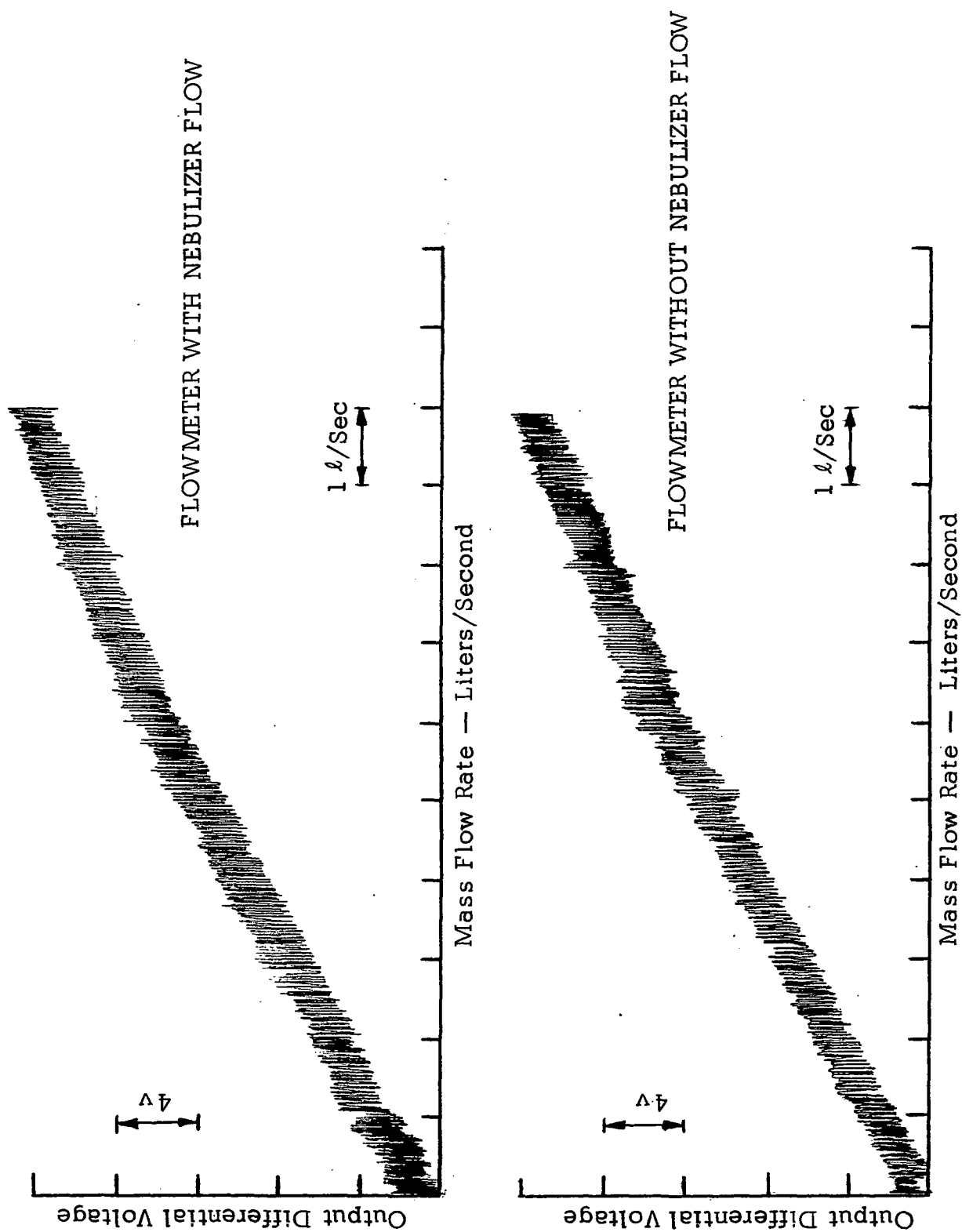


Figure 12. RESPIRATORY FLOWMETER VARIATION WITH NEBULIZER FLOW

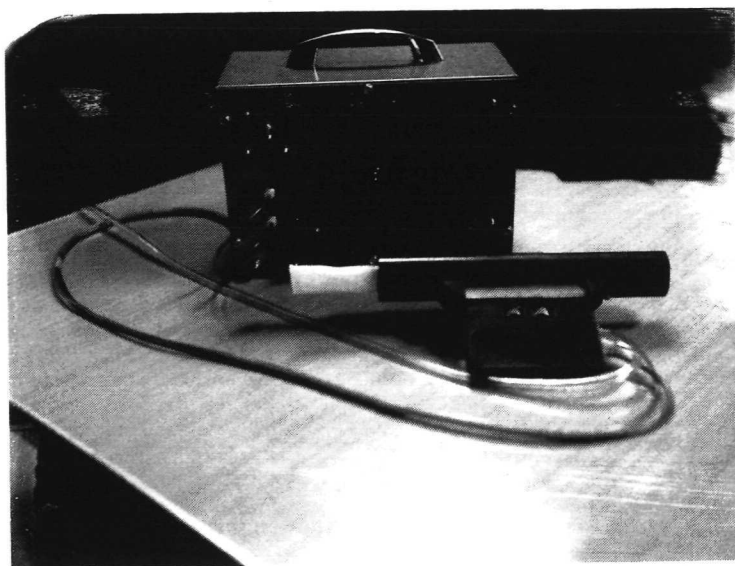


Figure 13. INTENSIVE CARE UNIT SYSTEM

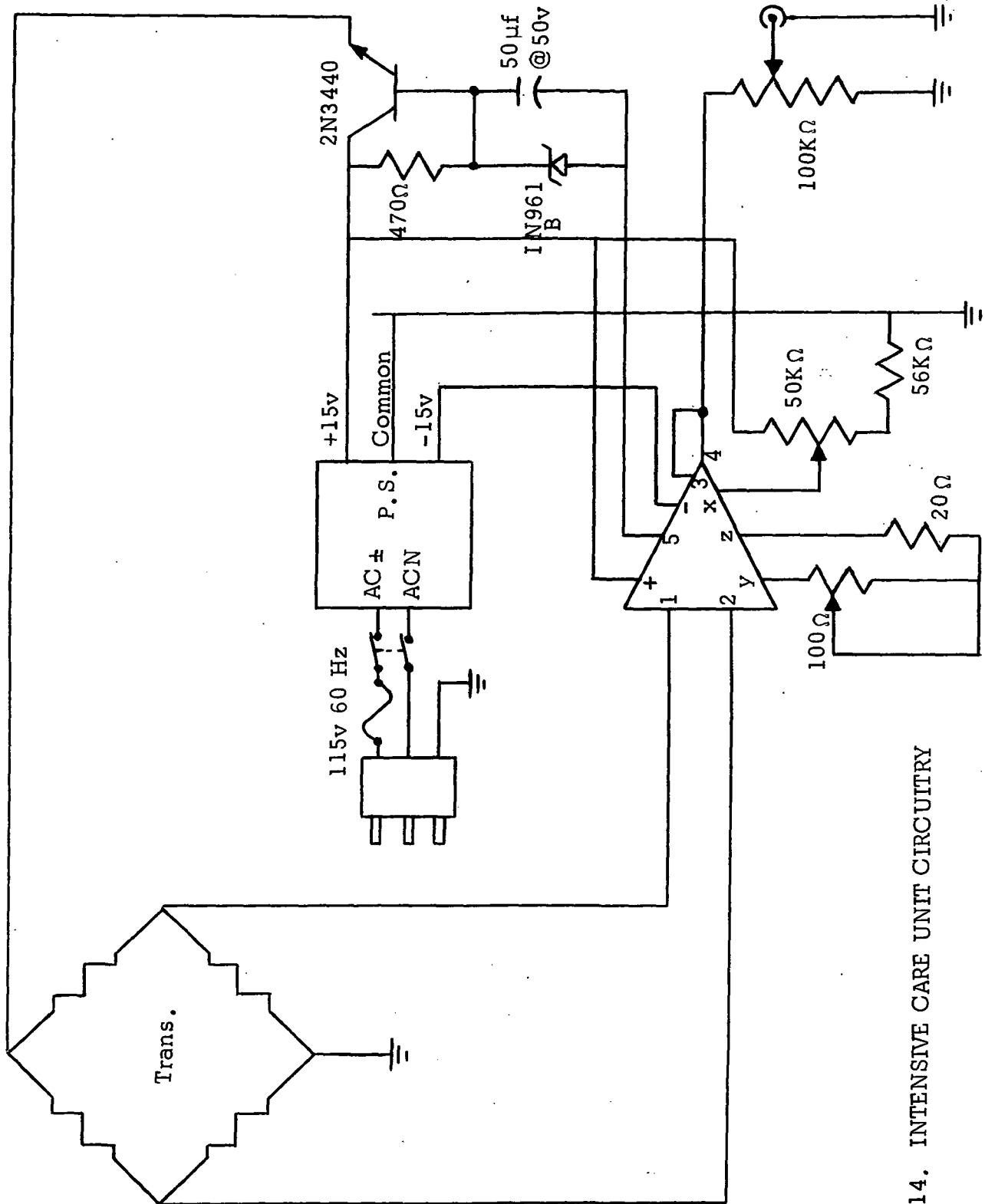
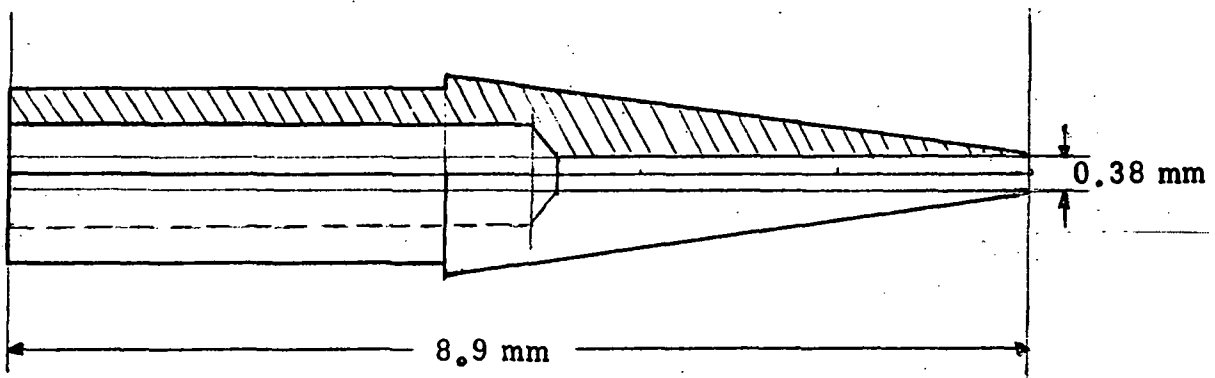


Figure 14. INTENSIVE CARE UNIT CIRCUITRY

dissolved. This process allows accurate formation of very small parts having thin wall sections (0.08 mm) impossible to obtain by other methods. The accuracy still depends on the ability to make duplicate parts, but these are solid (i.e., machinable) parts, the contour of which becomes the interior of the final product. A comparison of the output of the two nozzle types is shown in Figure 15. All subsequent tests were conducted using the circular nozzles.

One final aspect of the receiver design was modified; that is, the receiver was streamlined by the installation of conical inserts in the receiver orifice. A sketch of a nozzle insert is shown below.



The results of the test are shown in Figure 16. The noise has been reduced by a factor of 3, but the gain has been decreased by a factor of 1/3. The fact that the noise decreased is expected because of the RC filter characteristic of the receiver orifice. A surprising feature of the characteristic is the reduced gain. This may, however, be attributed to a misalignment of the receiver in the axis parallel to the flow.

B) Sensor geometry -- A variable geometry test fixture was fabricated to permit testing with adjustable separation and alignment. The fixture allowed nozzle adjustment perpendicular to the axis of flow by means of a micrometer. Separation between receiver and power nozzle was achieved by the use of gage blocks. The test fixture is shown in Figure 17.

A series of tests were run in an attempt to determine the optimum power nozzle to receiver nozzle spacing. The data is shown in Figure 18. Similarly, Figure 19 shows the results of tests where the receiver nozzles were held on the centerline of the flow constraining tube and the power nozzle was moved off-centerline.

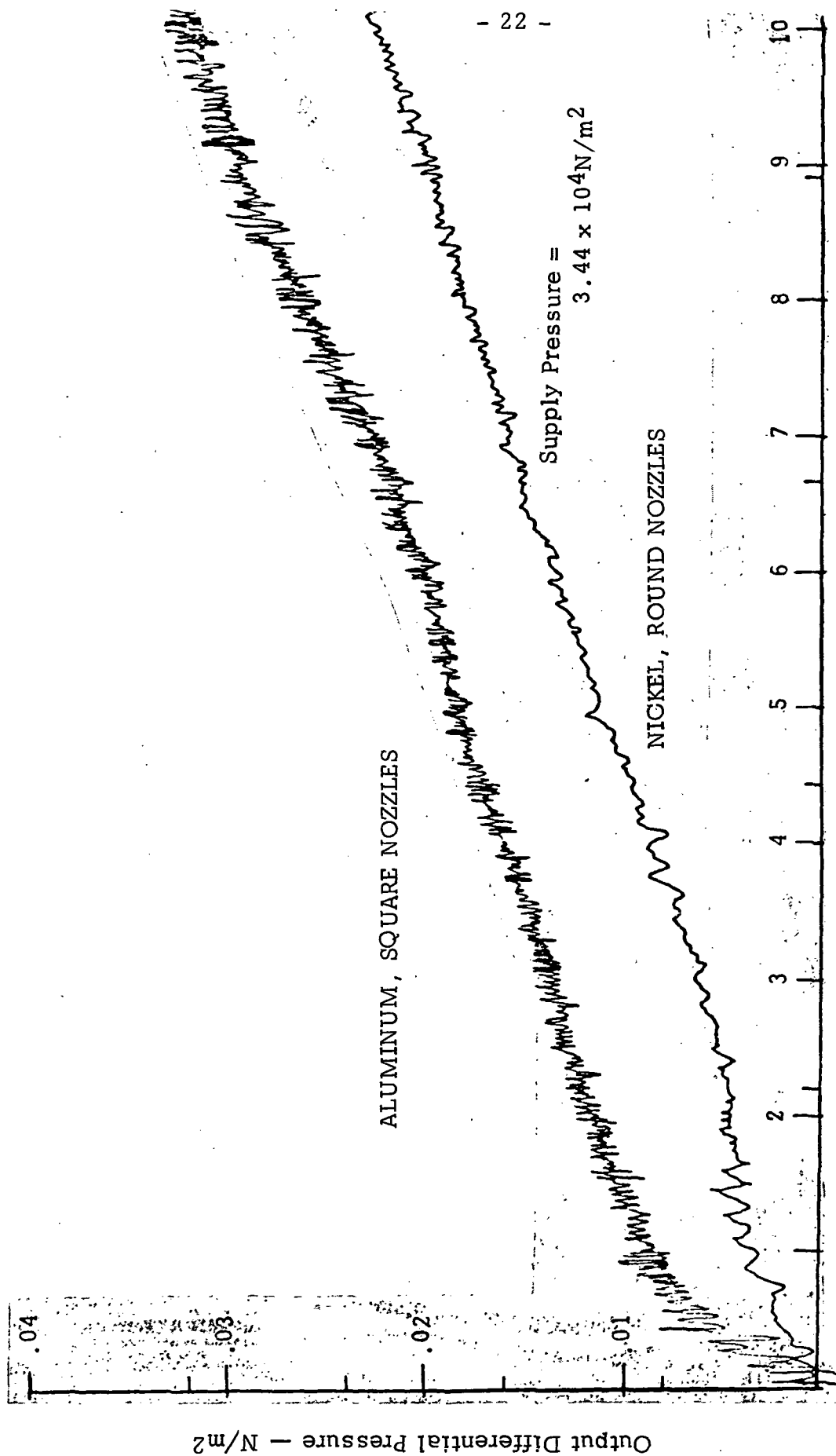


Figure 15. COMPARISON OF SQUARE NOZZLE SENSOR  
VERSUS ROUND NOZZLE SENSOR

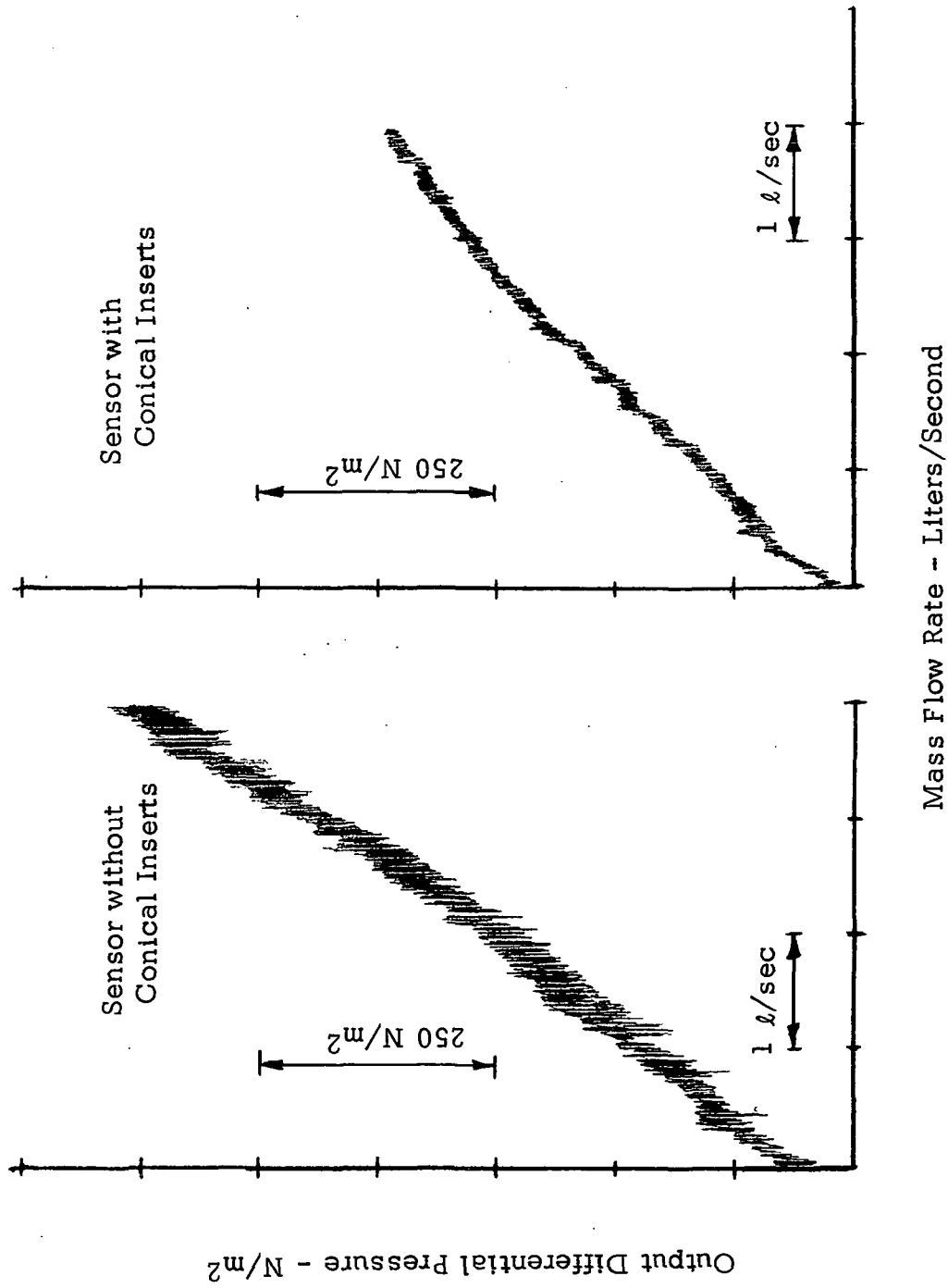


Figure 16. OUTPUT CHANGE WITH CONICAL INSERTS  
IN RECEIVER NOZZLES

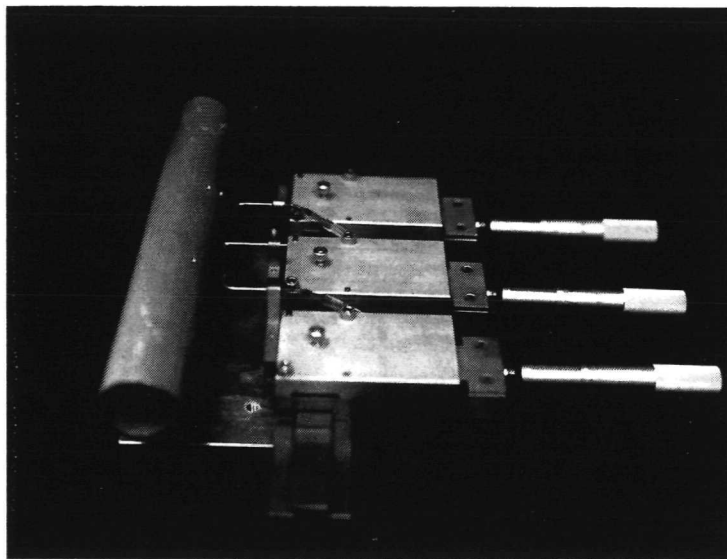


Figure 17.  
VARIABLE GEOMETRY TEST FIXTURE

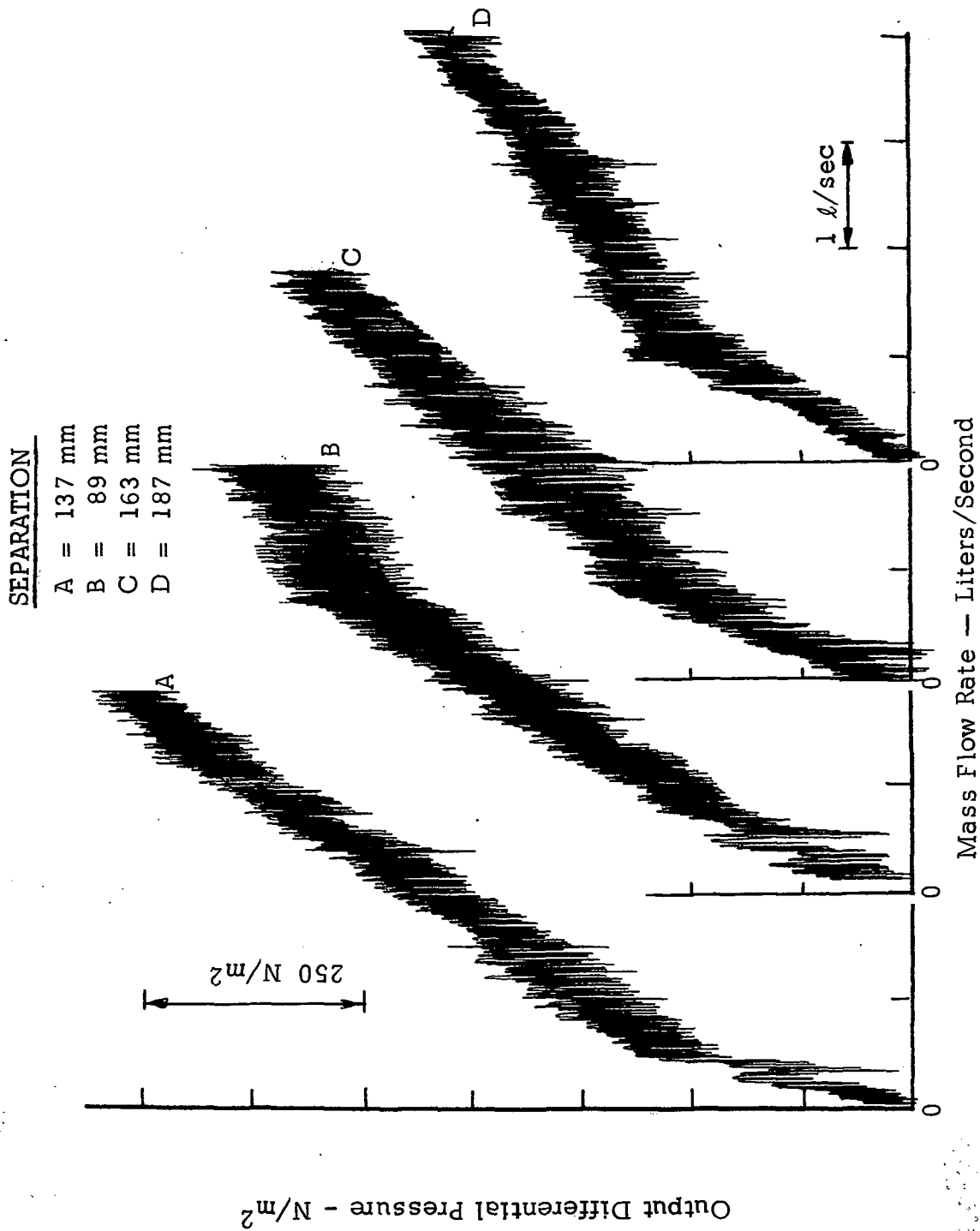


Figure 18. POWER NOZZLE - RECEIVER SEPARATION



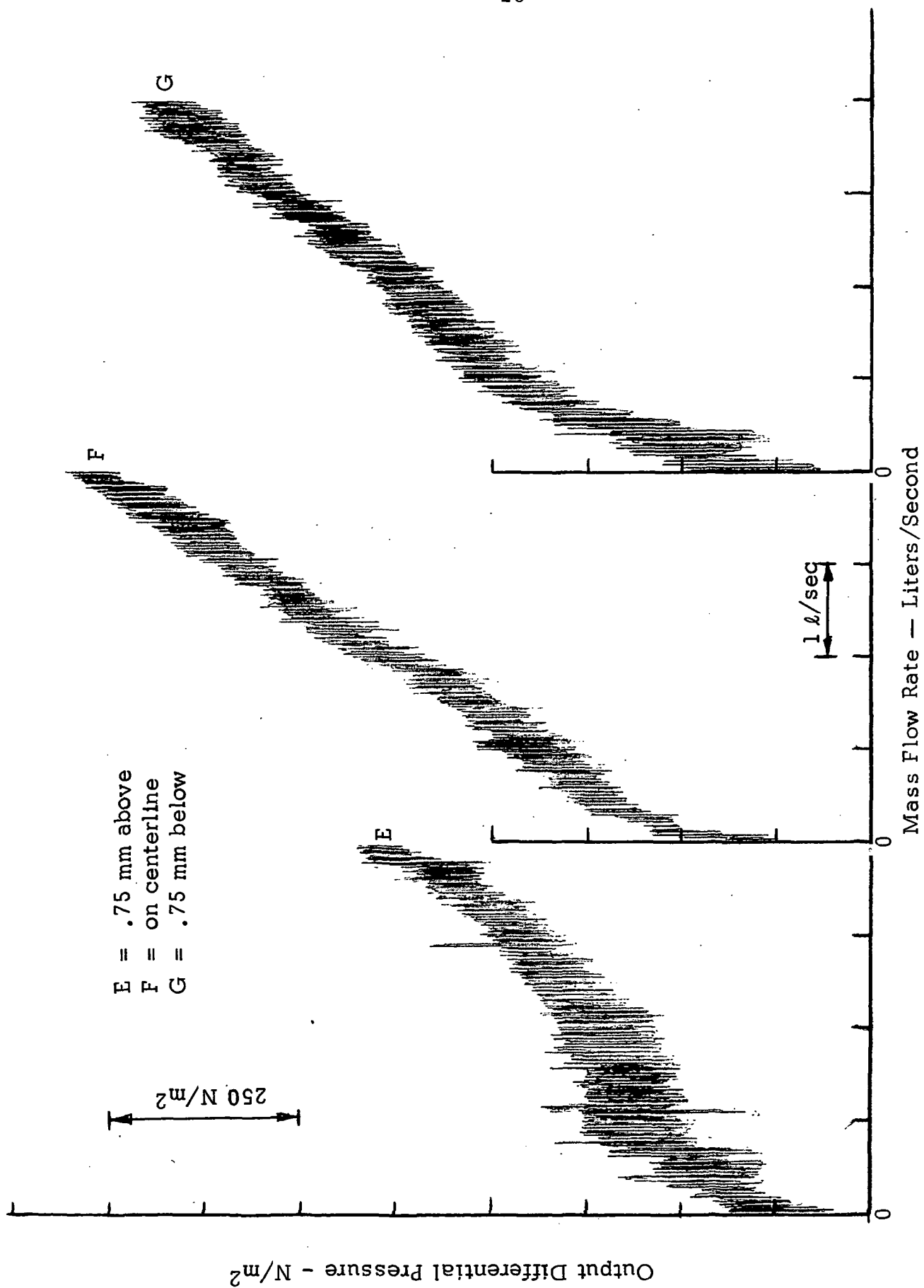


Figure 19. POWER NOZZLE - RECEIVER ALIGNMENT

This data is summarized in Figure 20a and 20b where the figure of merit is the signal-to-noise ratio defined as  $\frac{\Delta P/Q}{\Delta P_n}$ . Figure 20a indicates the highest signal-to-noise ratio occurred with a power nozzle-receiver separation of 137 mm. Figure 20b indicates the highest signal-to-noise ratio occurred with the power nozzle and receiver nozzles on the same centerline.

C) Flow confining tube diameter -- Since the velocity of the sensed flow is inversely proportional to the diameter squared, increased sensitivity results for a given flow rate when the diameter of the flow confining tube is reduced. Tests were conducted at confining tube diameters of 25 mm and 19 mm. The data obtained supports the assumption that the fluidically measured velocity does increase, yielding a corresponding increase in gain with a reduction in tube diameter. Of particular note is the fact that the noise is not significantly increased with such a reduction.

Opposing the resultant increased signal-to-noise ratio with reduction in tube diameter is the fact that the break point in the characteristic curve occurs at a lower flow rate. However, since the required flow rate of the Intensive Care Unit Flowmeter is less than one-half the required flow of the Pulmonary Function Unit (4 liters per second versus 10 liters per second), a reduction in the diameter from 25 mm to 19 mm was felt to offer a significant advantage.

D) Supply Pressure Variation -- It is known that the noise decreases with decreasing supply pressure. It is also known that the sensitivity decreases according to the square root of supply pressure. Whether or not a supply pressure reduction provides a quieter system depends on the noise decreasing at a rate greater than the square root of supply pressure. Experience has shown that the best performance for the Intensive Care Unit occurs at a supply pressure of  $20.7 \times 10^3 \text{ N/m}^2$ . This is in part due to the sensitivity of the electronics associated with the Intensive Care Unit.

### Calibration

The unit underwent preliminary calibration using the Hastings-Raydist linear mass flow meter. A steady-state calibration is shown in Figure 21.

In the interest of the final application of the fluidic respiratory flowmeter,

Figure 20a.

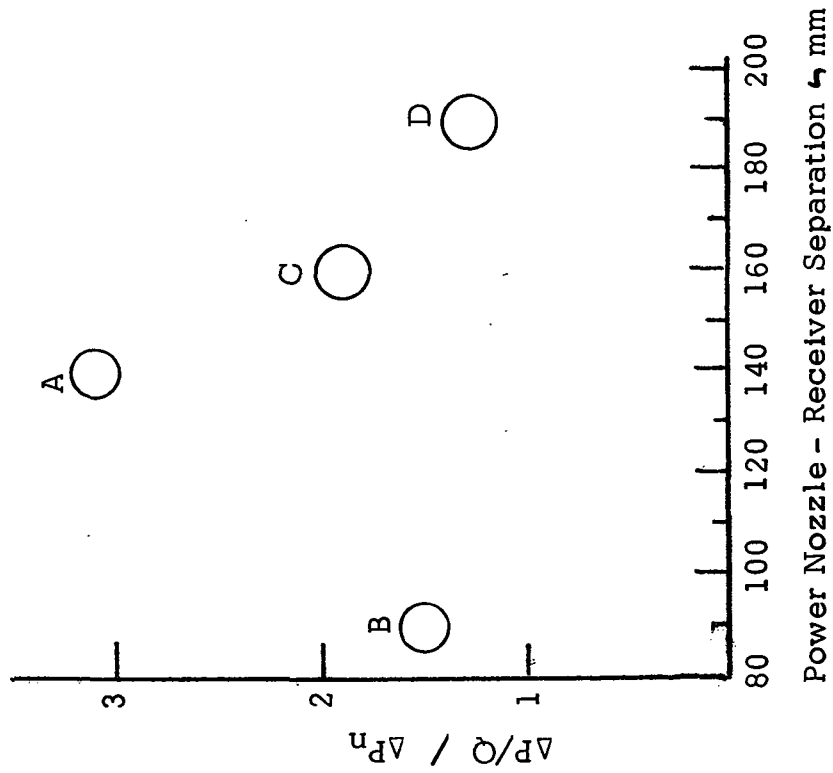


Figure 20b.

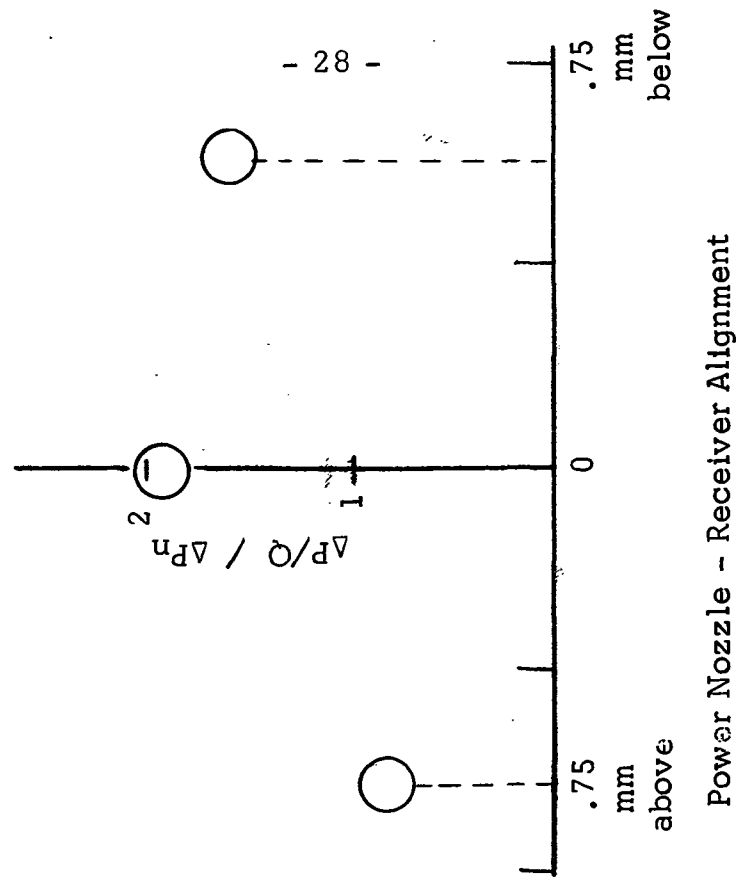


Figure 20a. -- Signal to Noise Ratio versus Power Nozzle-Receiver Separation  
Figure 20b. -- Signal to Noise Ratio versus Power Nozzle-Receiver Alignment

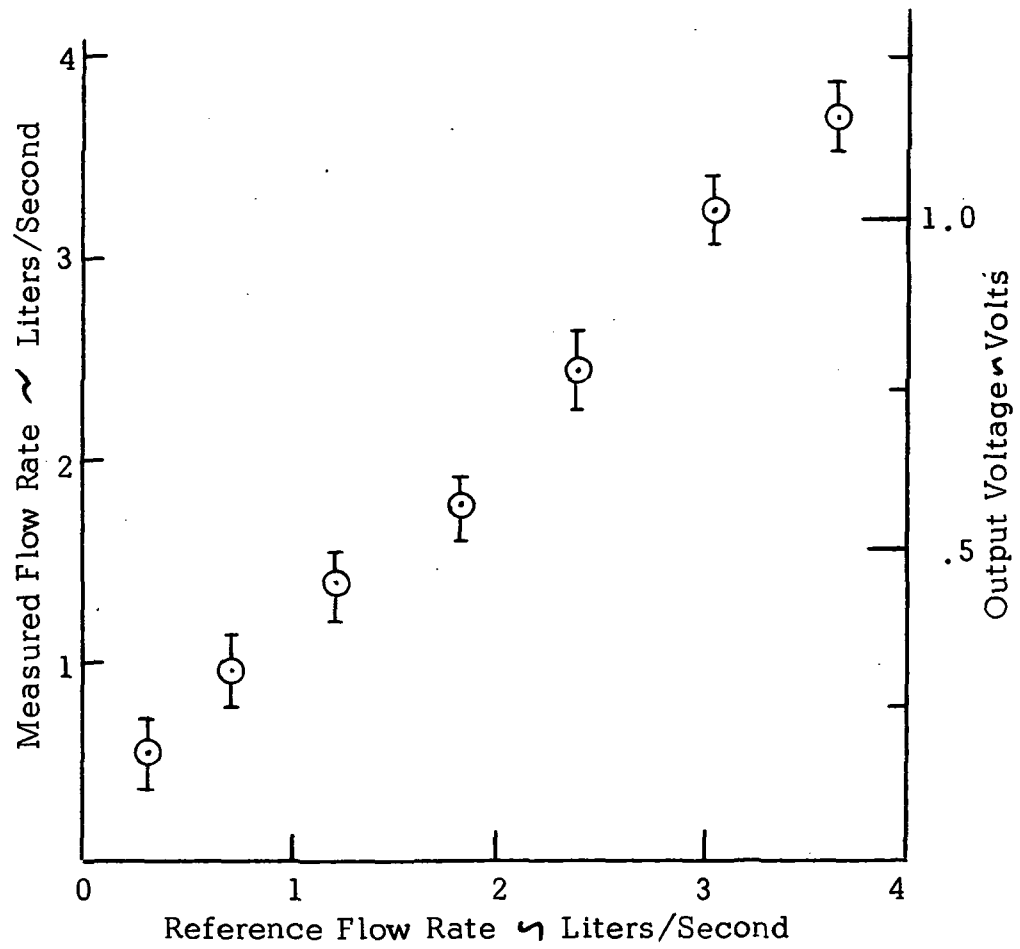


Figure 21. INTENSIVE CARE UNIT STEADY STATE CALIBRATION

the unit was calibrated against a Wedge Spirometer at the Georgetown University Hospital as was the Pulmonary Function Test Unit.

An exhalation trace was simultaneously plotted by both the Wedge Spirometer and by the Intensive Care Unit. The results are shown in Figures 22 and 23. Figure 22 compares the flow rate output trace of the Wedge Spirometer with that of the Intensive Care Unit. The Intensive Care Unit has a significantly higher noise level.

Although all of the techniques for reducing the sensor threshold which were derived in this program, were incorporated in final test unit, it was found that the degree of noise reduction demonstrated by the variable geometry fixture was not obtained in the final test unit. The fixed nature of the final sensor unit does not permit the necessary precise alignment that is inherently available in the variable geometry fixture. The incorporation of a precision alignment feature would unduly complicate the design and would constitute a departure for the program goal of simplicity.

Figure 23 shows the trace produced by the Wedge Spirometer compared with data measured from a trace of the Intensive Care Unit, i.e., the data was obtained by the use of polar planimeter techniques. The correlation is good except in the low flow region areas where fluidic noise again becomes a problem. There is fairly good correlation in the overall output characteristic.

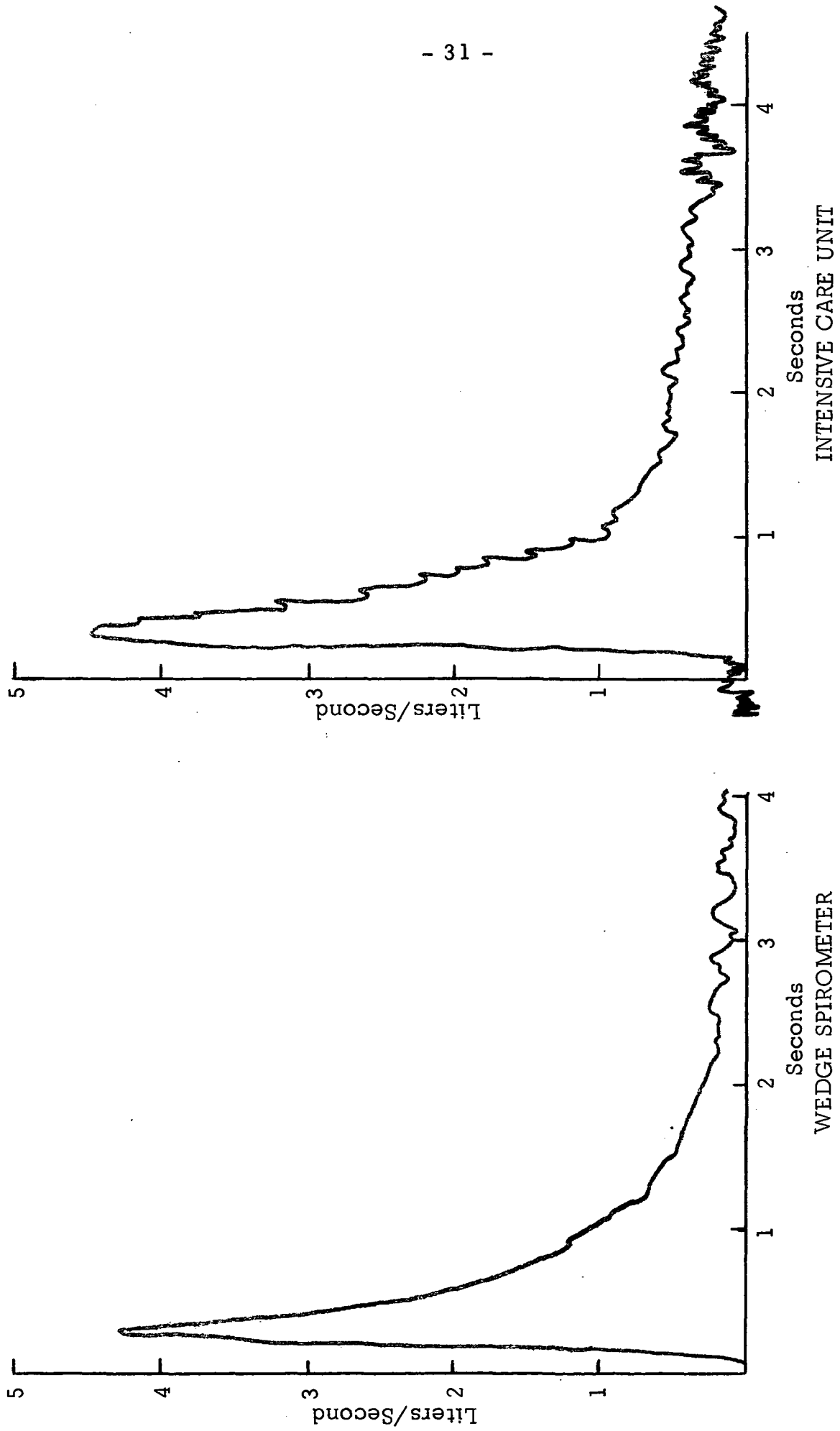


Figure 22.

EXHALATION TRACE - INTENSIVE CARE UNIT VERSUS WEDGE SPIROMETER FLOW RATE

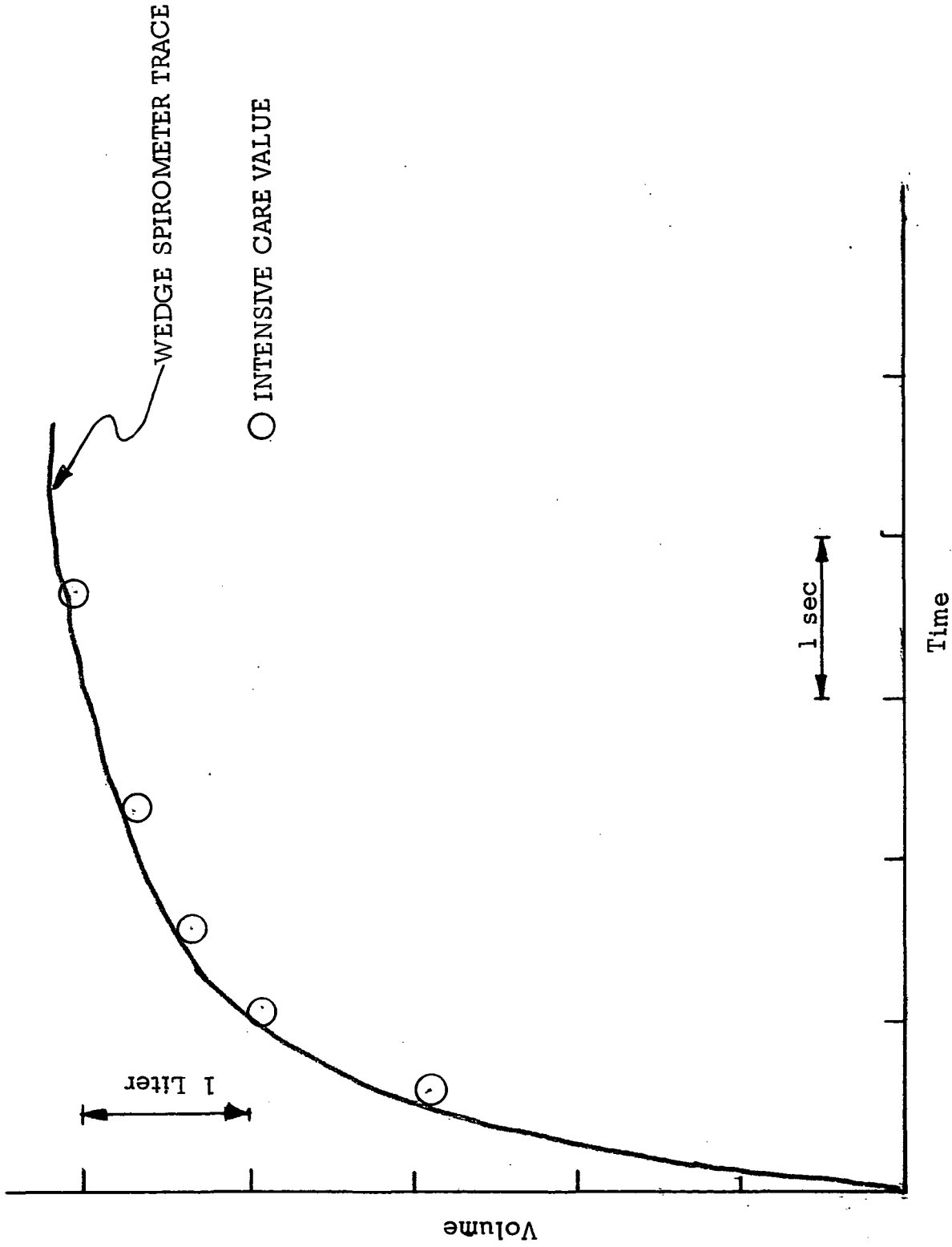


Figure 23.  
EXHALATION TRACE  
INTENSIVE CARE UNIT versus WEDGE SPIROMETER TOTAL VOLUME.

## SECTION 5.0

### SUMMARY

The Pulmonary Function Test Unit, as developed in this program, meets all of the program goals. The unit is fully self-contained with the exception of the air supply. It is felt that the greatest application for such a device is in the area of mass screening of patients. When lower threshold flow rates are required to quantify the patient's condition, mass screening of the type envisioned here may be accomplished simply by observation by medical personnel. That is when the patient is not capable of exhaling at a flow rate greater than .5 liters per second, his reduced capacity should readily be apparent. Such an individual requires more sophisticated equipment to quantify his ailment.

Whereas the low threshold requirement for the Pulmonary Function Test Flowmeter may be unnecessary, this is not the case for the Intensive Care Unit. This unit required a two order of magnitude reduction in noise to meet the program goal of 5 milliliters per second threshold. The reduction in noise brought about by this programs' investigation was approximately by a factor of 4.

Further noise reduction may be possible by the application of more sophisticated techniques but this would constitute a departure from the basic simplicity of the sensor. In conclusion, it may be feasible to adapt the aircraft air speed wind sensor to screen for Pulmonary Function disorders but it is not feasible to adapt the sensor to the more rigorous requirements of an Intensive Care Unit.



## SECTION 6.0

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13. ABSTRACT <p>A study was undertaken to determine the feasibility of adapting a fluidic airspeed sensor for use as a respiratory flowmeter. .</p> <p>A Pulmonary Function Testing Flowmeter was developed which should prove useful for mass screening applications. .</p> <p>The fluidic sensor threshold level was not reduced sufficiently to permit its adaptation to measuring the low respiratory flow rates encountered in many respiratory disorders. .</p>			

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